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CONVERSION OF CARDIAC PERFORMANCE  
DATA IN ANALOG FORM  
FOR DIGITAL COMPUTER ENTRY

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16. Abstract <p>A system is presented which will reduce analog cardiac performance data and convert the results to digital form for direct entry into a commercial time-shared computer. Circuits are discussed which perform the measurement and digital conversion of instantaneous systolic and diastolic parameters from the analog blood pressure waveform. Digital averaging over a selected number of heart cycles is performed on these measurements, as well as those of flow and heart rate. The determination of average cardiac output and peripheral resistance, including trends, is the end result after processing by digital computer.</p>					
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# CONVERSION OF CARDIAC PERFORMANCE DATA IN ANALOG FORM FOR DIGITAL COMPUTER ENTRY

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## SUMMARY

A method is presented for the automatic reduction of cardiac performance data in analog form using digital techniques. Conversion to digital paper tape format for entry into a commercial time-shared computer is also described.

A signal analysis technique is described which permits detection of systolic and diastolic pressure during each cardiac cycle. The general extrema measurement problem is discussed along with circuits which will accomplish the extrema measurement and provide outputs in digital form. An average of these peak measurements is accumulated digitally over a selectable number of heartbeat cycles and recorded on punched paper tape. Other parameters measured and recorded are heart rate, aortic blood flow, and integrated flow.

## INTRODUCTION

Most experimental research, regardless of the field, results in the need to gather and process large quantities of data. The technology developed at the NASA Lewis Research Center over a number of years in the measurement, recording, and processing of research data from propulsion systems testing has been found to be applicable to other fields. This was the case in a cooperative effort with the Research Division of Cleveland Clinic.

Medical research experiments were being conducted at the Clinic on the relation between cardiac output and renal hypertension in trained, unanesthetized dogs (refs. 1 and 2). Each animal was instrumented previously with a flow transducer implanted around the ascending aorta and an external pressure transducer linked by indwelling catheter to the femoral artery. The electromagnetic flowmeter signals were brought out through a skin connector at the back of the neck. The transducer outputs, following signal conditioning, were recorded on analog magnetic tape for later processing. Fig-

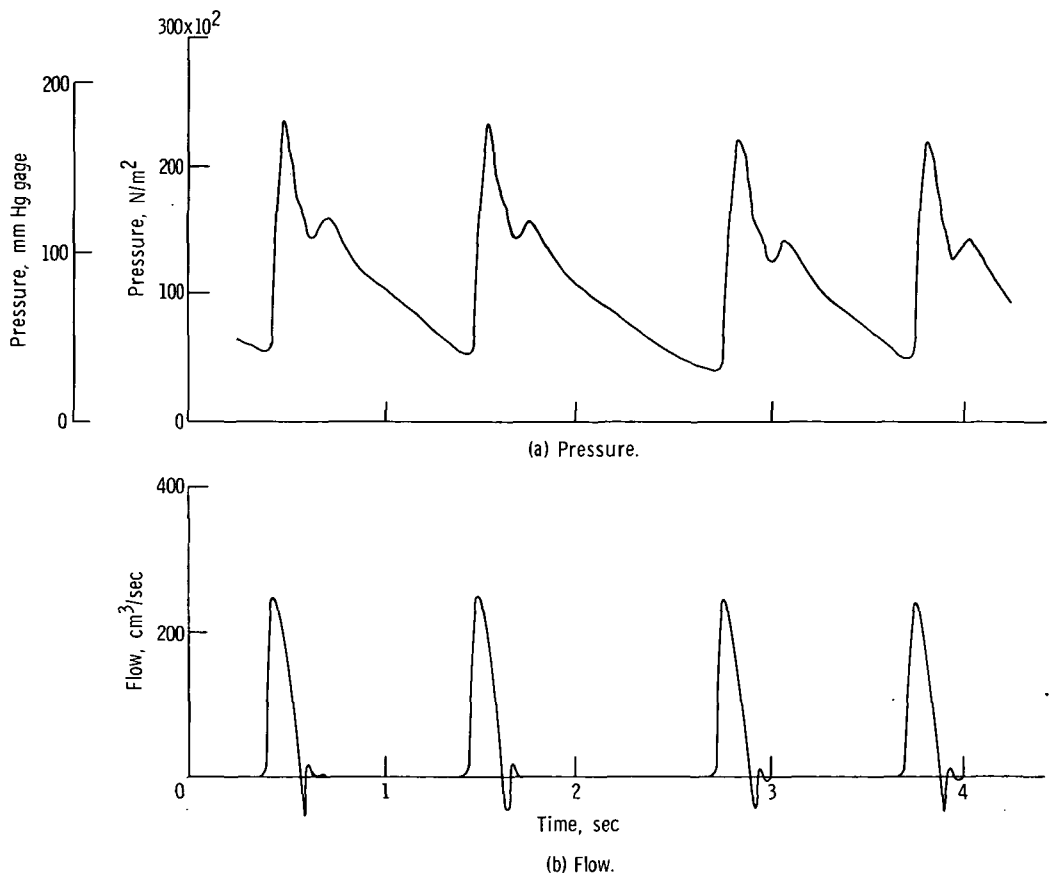


Figure 1. - Typical blood pressure and flow waveforms.

Figure 1 shows the typical blood pressure and blood flow waveforms as they were recorded originally in one dog.

Data were collected over a large number of tests with many different animals. The original hand analysis technique consisted of making analog strip charts of the pressure and flow waveforms from the analog tape. The instantaneous diastolic and systolic pressure parameters were read manually from the strip charts and averaged over a number of samples. The blood flow peak and width were measured. The blood flow waveform was also integrated electronically and recorded on the strip chart. This integrated blood flow was calibrated to represent the area under the blood flow waveform, which is the stroke volume. These measurements were also averaged over a number of samples, as was heart rate. Finally, the measurements were used to determine both average cardiac output and total peripheral resistance during the test period.

Data were collected for several minutes on each animal once or twice each day for up to 4 months. The data volume quickly created a monumental hand analysis task. Estimates approached 1 man year full-time, and additional similar experiments would have compounded the problem.

The Lewis Research Center was asked to look at the analysis problem and determine whether technology developed in aerospace applications might be applied to this experiment. Specifically, a system was needed which would accept the pressure and flow signals in analog form and process these for entry into a commercial time-shared digital computer for trend analysis.

This report describes the analysis of the blood pressure analog waveform, describes the circuits used to convert the measurements of interest to digital form, and shows how these circuits were incorporated into a system. The detailed system is not described in this report.

## WAVEFORM ANALYSIS

The specific parameters of interest in the experiment were the systolic (maximum peak) and diastolic (minimum peak) blood pressure, peak blood flow, flow pulse width (in time), and heart rate. Since the most difficult electronic measurement problem is in the peak measurements, the blood pressure waveform analysis is considered first (see fig. 1(a)).

The baseline shown represents atmospheric pressure to which the transducer is referenced. The aortic blood pressure waveform falls within the range of  $53.2 \times 10^2$  to  $266 \times 10^2$  newtons per square meter (40 to 200 mm Hg). The electrical analog of this pressure waveform falls into the range of zero to 2.5 volts, where zero represents atmospheric pressure. This voltage appears at the output of the signal conditioner connected to the transducer and also at the output of the analog recorder on playback. This is the signal which must be analyzed by electronic circuits and converted to digital form.

Measurement of the voltages representing the systolic and diastolic pressures presents a special problem. The conventional circuits measure the maximum excursion from a baseline but do not measure the minimum (ref. 3).

Several attempts were made to shift the baseline to some center point in the waveform so that the conventional peak measuring circuits could be employed. The intent was to then measure systolic as the maximum positive excursion from the baseline and diastolic as the maximum negative excursion from this baseline.

It was determined that no baseline could be chosen that guaranteed a baseline crossing on every cardiac cycle. Figure 2 shows this problem, in this case resulting in a loss of one diastolic measurement.

One technique explored permitted the waveform to "float" between two baselines which bracket peak positive and negative limits (see fig. 3).

The only constraint in the selection of the two baselines is that the positive peak must always be above its reference and the negative peak must always be below its reference. This double baseline technique avoids the baseline crossing problem but intro-

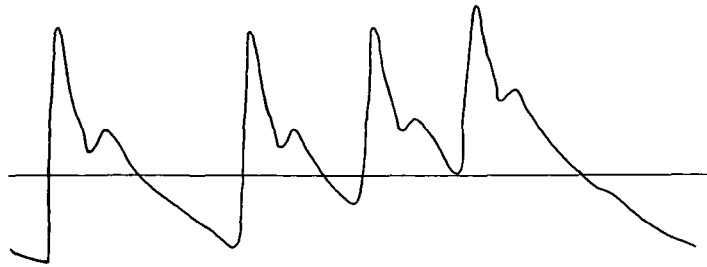


Figure 2. - Pressure trace with center baseline drawn in.

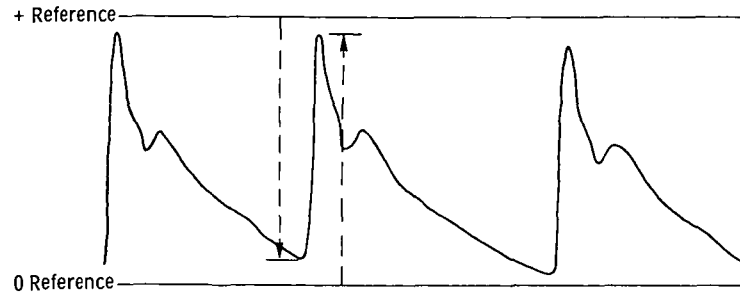


Figure 3. - Pressure trace with double baseline.

duces another one. It is desirable to be able to compare raw data measurements of diastolic and systolic pressures for preliminary analysis. This comparison is difficult if the measurements are made from different references. In order to compare the two pressures, the diastolic measurement must be subtracted from the difference between the two references.

The technique finally used, as shown in figure 4, avoided this subtraction problem. The technique uses a single base reference line below the lowest diastolic pressure, corresponding to atmospheric pressure. The electrical reference in this case is ground. Coincidentally, this is the reference technique used in the physical measurement of blood pressure.

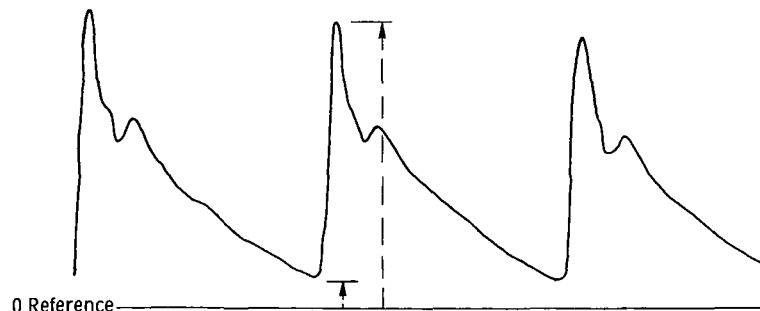


Figure 4. - Pressure trace with single low baseline.

## MEASUREMENT CIRCUITS

The maximum peak voltage measurement, representing the systolic blood pressure, uses the conventional peak measuring circuit, as shown in figure 5.

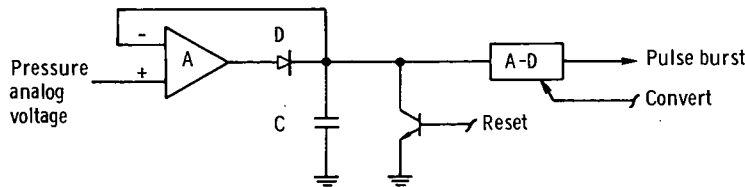


Figure 5. - Systolic measurement circuit.

An analog follower A, charges the capacitor C through a diode D to the most positive voltage in each cycle. At some fixed point in each cycle after the peak is reached, a convert signal commands the analog to digital converter (A-D) to measure the voltage stored on the capacitor C. It produces a number of pulses proportional to this voltage at its output. After the conversion is complete, the reset signal discharges the capacitor to the ground reference, and the cycle is repeated.

The diastolic measurement presents a special problem because it represents a minimum excursion from the ground reference baseline. The conventional circuit was modified by interchanging the roles of the follower-diode and reset (see fig. 6). The

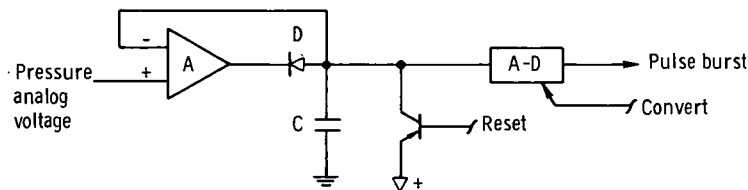


Figure 6. - Diastolic measurement circuit.

function of the reset circuit in this case is to charge the capacitor to a voltage higher than the highest voltage to be measured (full scale). The follower-diode then discharges the capacitor to the lowest peak in the cycle, after which the A-D converter measures that lowest voltage and produces a pulse burst proportional to it.

The accuracy of this measurement is not affected by the charging voltage, provided that it exceeds the highest minimum peak voltage to be measured. Note that, because the measurement is still made from the ground reference baseline, it is possible to compare the raw pulse counts representing diastolic and systolic measurements.

## THE SYSTEM

A system incorporating this single baseline measurement technique was built, part of which is shown in simplified form in figure 7.

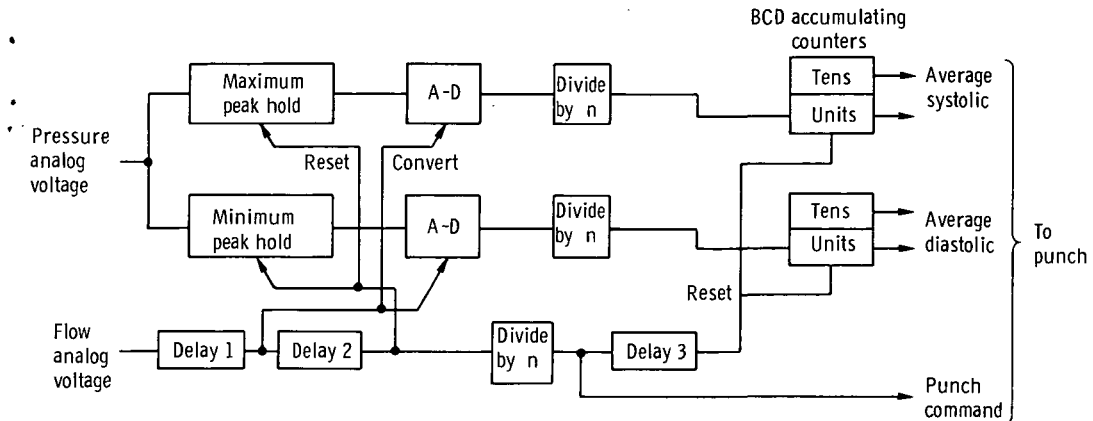


Figure 7. - Simplified system.

The blood pressure analog waveform is analyzed by the maximum and minimum peak holding circuits, as previously described. Some means had to be provided for the positive identification of a fixed time in each heartbeat cycle to trigger the holding and conversion circuits. Fortunately, in this particular application, the blood flow waveform was one of the parameters measured by a separate transducer. Its waveform is a very sharp, clean pulse and does not change significantly in shape throughout an experiment. The leading edge of this pulse is at the start of systole. If this waveform had not been available, a technique for determining the beginning of systole by using the pressure waveform alone as described in reference 4 could have been used.

In the partial system diagram, figure 7, the two input signals are shown on the left side. The pressure analog voltage is analyzed simultaneously by the minimum and maximum peak holding circuits. At the start of systole, which is the very leading edge of the flow waveform, delay 1 is started. The time period for delay 1 is about 200 milliseconds, which is long enough for the systolic pressure peak to be reached. Both peak holding circuits are fully charged at this time since the diastolic pressure peak is reached at the start of systole. The A-D converter is now commanded to convert both channels to a burst of pulses.

After the convert command occurs at the end of delay 1, delay 2 is initiated. Delay 2 has a duration of about 10 milliseconds, which allows time for the A-D converters to produce a pulse stream. At the end of delay 2, a momentary reset is generated which resets



the holding circuits for the next cycle.

A typical pulse burst on a single systolic measurement would contain 80 pulses. If we assume that the three "Divide by  $n$ " boxes are set for  $n = 10$ , the binary coded decimal (BCD) accumulating counter would contain 8 after 1 pulse burst and 80 after 10 bursts. At the end of 10 bursts, the punch is commanded to punch the contents of the two counters containing the average peak measurements, and delay 3 resets the counters. The cycle is then repeated. The number  $n$  may be set to 10, 20, or 40 beats in the actual system built.

The flow peak, flow period, and heart rate are measured by conventional circuits and therefore are not described. These are also punched by the paper tape punch as an average over the same  $n$  samples.

## CONCLUDING REMARKS

The system presented in this report is intended to be used for the preparation of analog recorded cardiac performance parameters for entry into a digital computer. However, the analysis technique and circuits could easily be used to provide digital readout for on-line patient monitoring. Entry into a digital computer for the monitoring of multiple patients, as required in intensive care wards, is a possible application.

It was assumed, in this report, that the calibration steps are performed by the digital computer. The raw data numbers punched into the paper tape, which have a range of 0 to 99 for each parameter, must be multiplied by the proper calibration coefficients to represent pressure, flow, and rates. In a patient monitoring system which does not have a computer, but does need a digital readout, analog circuits could be employed ahead of the analog-to-digital conversion that would perform the calibration function.

The system described was built under contract and is being put into service at the Research Division of Cleveland Clinic in studies related to hypertension. In addition to the measurements described, flow peak, flow width, and integrated flow are recorded, along with the time period required for  $n$  beats. All parameters are punched by a high-speed paper tape punch in a format suitable for entry into a commercial time-shared computer terminal. Consequently, the collection and processing of data can be totally automated.

Lewis Research Center,  
National Aeronautics and Space Administration,  
Cleveland, Ohio, December 22, 1971,  
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